Computational Methods in Applied Sciences

Volume 53

Series Editor

Eugenio Oñate, Universitat Politècnica de Catalunya, Barcelona, Spain
This series publishes monographs and carefully edited books inspired by the thematic conferences of ECCOMAS, the European Committee on Computational Methods in Applied Sciences. As a consequence, these volumes cover the fields of Mathematical and Computational Methods and Modelling and their applications to major areas such as Fluid Dynamics, Structural Mechanics, Semiconductor Modelling, Electromagnetics and CAD/CAM. Multidisciplinary applications of these fields to critical societal and technological problems encountered in sectors like Aerospace, Car and Ship Industry, Electronics, Energy, Finance, Chemistry, Medicine, Biosciences, Environmental sciences are of particular interest. The intent is to exchange information and to promote the transfer between the research community and industry consistent with the development and applications of computational methods in science and technology.

Book proposals are welcome at
Eugenio Oñate
International Center for Numerical Methods in Engineering (CIMNE)
Technical University of Catalunya (UPC)
Edificio C-1, Campus Norte UPC Gran Capitán
s/n08034 Barcelona, Spain
onate@cimne.upc.eduwww.cimne.com
or contact the publisher, Dr. Mayra Castro, mayra.castro@springer.com

Indexed in SCOPUS, Google Scholar and SpringerLink.

More information about this series at http://www.springer.com/series/6899
Multibody Dynamics 2019

Proceedings of the 9th ECCOMAS Thematic Conference on Multibody Dynamics

Springer
Preface

The present proceedings book collects a selection of full-papers as a subset of the submitted extended abstracts presented at the 9th ECCOMAS Thematic Conference on Multibody Dynamics, held on 15–18 July 2019, at the University of Duisburg-Essen, Duisburg, Germany.

Multibody dynamics is an exciting area of mechanics which merges various disciplines such as structural dynamics, multiphysics problems, computational mathematics, control theory and computer science in order to deliver methods and tools for the virtual prototyping of complex mechanical systems. In this setting, multibody dynamics play a central role in the modelling, analysis, simulation, and optimization of mechanical systems in a variety of fields and for a wide range of applications. As new methods and procedures are being proposed at a fast pace in academia, research laboratories and industry, it becomes important to provide researchers in multibody dynamics with appropriate venues for exchanging ideas and results. To answer these needs, the ECCOMAS Thematic Conference on Multibody Dynamics was initiated in 2003 in Lisbon. Since this event, the conference was held every two years in changing locations and with large attraction in the international scientific community: Madrid (2005), Milan (2007), Warsaw (2009), Brussels (2011), Zagreb (2013), Barcelona (2015), and Prague (2017). The objective of this conference series was to bring together researchers from different fields where multibody dynamics play a key role. This includes not only theoretical fields where multibody dynamics were traditionally established, but also in particular applications in which multibody dynamics might contribute new perspectives for new-generation practical and industrial challenges.

The present conference has attracted again a strong interest in the international community: a total of 165 contributions from 27 countries were selected for presentation after severe peer review by two independent reviewers. From the submitted full papers, 64 articles were anew selected after severe peer review by two independent reviewers. This book contains the thus identified full papers, which reflect the importance multibody dynamics research has in current and future multibody dynamics technologies from technical systems over robotic control up to human motion interactions.
The book is of interest to researchers, doctoral students, teachers and engineers specializing in multibody dynamics. It is divided into eight sections corresponding to the wide range of methodologies and applications encompassed in this thriving field: biomechanics; contact and constraints; mechatronics, robotics and control; flexible multibody dynamics; formulations and numerical methods; optimization and sensitivity analysis; efficient simulation and real-time applications; and applications in vehicle dynamics and aerospace devices.

We thank the authors for submitting their valuable contributions for this conference as well as the reviewers for performing the reviews in due time. We also thank the publisher Springer for the timely implementation of this book and the valuable advice during the production process. We are very indebted to the University of Duisburg-Essen as well as a long list of sponsors for the operational and financial support of this conference. Last but not least, we thank the European Community on Computational Methods in Applied Sciences, ECCOMAS, for the ideal support by offering its patronage for this conference.

Andrés Kecskeméthy
Francisco Geu Flores
## Contents

### Biomechanics

**Investigation of Tympanic Membrane Influences on Middle-Ear Impedance Measurements and Simulations**

Benjamin Sackmann, Birthe Warnholtz, Jae Hoon Sim, Dmitrii Burovikhin, Ernst Dalhoff, Peter Eberhard, and Michael Lauxmann

---

**Anterior Cruciate Ligament Injuries Alter the Kinematics and Kinetics of Knees with or Without Meniscal Deficiency**

Xiaode Liu, Hongshi Huang, Shuang Ren, Yingfang Ao, and Qiguo Rong

---

**Investigation of Inhomogeneous Stiffness and Damping Characteristics of the Human Stapedial Annular Ligament**

D. Burovikhin, Benjamin Sackmann, Merlin Schär, J. H. Sim, P. Eberhard, and M. Lauxmann

---

**Comparison of Measured EMG Data with Simulated Muscle Actuations of a Biomechanical Human Arm Model in an Optimal Control Framework – Direct Vs. Muscle Synergy Actuation**

Marius Obentheuer, Michael Roller, Staffan Björkenstam, Karsten Berns, and Joachim Linn

---

**A Detailed Kinematic Multibody Model of the Shoulder Complex After Total Shoulder Replacement**

Sven Herrmann, Mårnan Kebbach, Robert Grawe, Kelsey Kubiak, Katrin Ingr, Rainer Bader, and Christoph Woernle

---

**Multibody Analysis of a 3D Human Model with Trunk Exoskeleton for Industrial Applications**

Elisa Panero, Giovanni Gerardo Muscolo, Laura Gastaldi, and Stefano Pastorelli
# Contents

A Hill Muscle Actuated Arm Model with Dynamic Muscle Paths ........ 52
Johann Penner and Sigrid Leyendecker

Optimal Control Simulations of Two-Finger Precision Grasps ............ 60
Uday Phutane, Michael Roller, Staffan Björkenstam, and Sigrid Leyendecker

Reinforcement Learning Applied to a Human Arm Model ............... 68
Michael Burger, Simon Gottschalk, and Michael Roller

**Contact and Constraints**

Dynamic Modeling and Analysis of Pool Balls Interaction ............... 79
Eduardo Corral, Raúl Gismeros, Filipe Marques, Paulo Flores, Maria Jesús Gómez García, and Cristina Castejon

Dynamics of Machine-Process Combinations .......................... 87
Friedrich Pfeiffer

Modeling of Elastic Cages in the Rolling Bearing Multi-Body Tool CABA3D ......................................................... 96
Dmitry Vlasenko and Bodo Hahn

Analysis of the Influence of the Links’ Flexibility and Clearance Effects on the Dynamics of the RUSP Linkage ............... 104
Krzysztof Augustynek and Andrzej Urbaś

**Mechatronics, Robotics and Control**

Multibody Analysis and Design of an Electromechanical System Simulating Hyperelastic Membranes ............................. 115
Valentina Franchi, Gianpietro Di Rito, Roberto Galatolo, Ferdinando Cannella, Darwin Caldwell, and Giovanni Gerardo Muscolo

Haptic Simulation of Mechanisms ........................................ 123
Jascha Norman Paris, Jan-Lukas Archut, Mathias Hüsing, and Burkhard Corves

Solution Techniques for Problems of Inverse Dynamics of Flexible Underactuated Systems ............................. 131
Timo Ströhle and Peter Betsch

Investigation of the Behavior of Vibration-Damped Flexible Link Robots in End-Effector Contact: Simulation and Experiment ........ 139
Florian Pucher, Hubert Gattringer, and Andreas Müller

Possibilistic Investigation of Mechanical Control Systems Under Uncertainty ..................................................... 147
Andreas Hofmann, Michael Hanss, and Peter Eberhard
## Contents

<table>
<thead>
<tr>
<th>Title</th>
<th>Pages</th>
</tr>
</thead>
<tbody>
<tr>
<td>Nonlinear Position Control of a Very Flexible Parallel Robot Manipulator</td>
<td>155</td>
</tr>
<tr>
<td>Peter Eberhard and Fatemeh Ansarieshlaghi</td>
<td></td>
</tr>
<tr>
<td>A Compliant and Redundantly Actuated 2-DOF 3RRR PKM: Best of Both Worlds?</td>
<td>163</td>
</tr>
<tr>
<td>Robin Cornelissen, Andreas Müller, and Ronald Aarts</td>
<td></td>
</tr>
<tr>
<td>On the Modeling of Redundantly-Actuated Mechanical Systems</td>
<td>172</td>
</tr>
<tr>
<td>Yaojun Wang, Bruno Belzile, Jorge Angeles, and Qinchuan Li</td>
<td></td>
</tr>
<tr>
<td>An Individual Pitch Control Concept for Wind Turbines Based on Inertial Measurement Units</td>
<td>180</td>
</tr>
<tr>
<td>János Zierath, Thorben Kallen, Dirk Machost, Reik Bockhahn, Thomas Konrad, Sven-Erik Rosenow, Uwe Jassmann, and Dirk Abel</td>
<td></td>
</tr>
<tr>
<td>Flexible Multibody Dynamics</td>
<td></td>
</tr>
<tr>
<td>Localized Helix Configurations of Discrete Cosserat Rods</td>
<td>191</td>
</tr>
<tr>
<td>Vanessa Dörlich, Tomas Hermansson, and Joachim Linn</td>
<td></td>
</tr>
<tr>
<td>Importance of Warping in Beams with Narrow Rectangular Cross-Sections: An Analytical, Numerical and Experimental Flexible Cross-Hinge Case Study</td>
<td>199</td>
</tr>
<tr>
<td>Marijn Nijenhuis, Ben Jonker, and Dannis Brouwer</td>
<td></td>
</tr>
<tr>
<td>Robust and Fast Simulation of Flexible Flat Cables</td>
<td>207</td>
</tr>
<tr>
<td>Michael Roller, Christoffer Cromvik, and Joachim Linn</td>
<td></td>
</tr>
<tr>
<td>Dynamic Analysis of Compliant Mechanisms Using Absolute Nodal Coordinate Formulation and Geometrically Exact Beam Theory</td>
<td>215</td>
</tr>
<tr>
<td>Zhigang Zhang, Xiang Zhou, and Zhanpeng Fang</td>
<td></td>
</tr>
<tr>
<td>Dynamic Performance of Flexible Composite Structures with Dielectric Elastomer Actuators via Absolute Nodal Coordinate Formulation</td>
<td>223</td>
</tr>
<tr>
<td>Haidong Yu, Yunyong Li, Aolin Chen, and Hao Wang</td>
<td></td>
</tr>
<tr>
<td>Approaches to Fibre Modelling in the Model of an Experimental Laboratory Mechanical System</td>
<td>231</td>
</tr>
<tr>
<td>Pavel Polach, Michal Hajžman, and Radek Bulin</td>
<td></td>
</tr>
<tr>
<td>Body-Fluid-Structure Interaction Simulation for a Trailing-Edge Flexible Stabilizer</td>
<td>239</td>
</tr>
<tr>
<td>Abolfazl Kiani and Meisam Mohammadi-Amin</td>
<td></td>
</tr>
<tr>
<td>Investigation of a Model Update Technique for Flexible Multibody Simulation</td>
<td>247</td>
</tr>
<tr>
<td>Andreas Schulze, Johannes Luthe, János Zierath, and Christoph Woernle</td>
<td></td>
</tr>
</tbody>
</table>
Extension of the Iterative Improved Reduced System Technique to Flexible Mechanisms ........................................... 255
Alessandro Cammarata, Rosario Sinatra, and Pietro Davide Maddio

Updating of Finite Element Models for Controlled Multibody Flexible Systems Through Modal Analysis .......................... 264
Dario Richiedei and Alberto Trevisani

Formulations and Numerical Methods

Modelling Rigid and Flexible Bodies with Truss Elements ............. 275
Jacob Philippus Meijaard

State Observation in Beam-Like Structures Under Unknown Excitation ................................................................. 283
Johannes Luthe, Andreas Schulze, Roman Rachholz, János Zierath, and Christoph Woernle

Dynamic Modelling of Lower Mobility Parallel Manipulators .......... 292
Haitao Liu, Weifeng Chen, Tian Huang, Huafeng Ding, and Andres Kecskemethy

Mathematical Model of a Crane with Taking into Account Friction Phenomena in Actuators ........................................ 299
Andrzej Urbaś and Krzysztof Augustynek

Closed Form of the Baker-Campbell-Hausdorff Formula for the Lie Algebra of Rigid Body Displacements ...................... 307
Daniel Condurache and Ioan-Adrian Ciureanu

Alternative Integration Schemes for Constrained Mechanical Systems ................................................................. 315
Tobias Meyer, Pu Li, and Bernhard Schweizer

Implementation of Linear Springs and Dampers in a Newmark Second Order Direct Integration Method for 2D Multibody Dynamics ................................................................. 323
Haritz Uriarte, Igor Fernández de Bustos, and Gorka Urkullu

On the Numerical Treatment of Nonlinear Flexible Multibody Systems with the Use of Quasi-Newton Methods ............... 332
Radek Bulín and Michal Hajžman

Interior-Point Solver for Non-smooth Multi-Body Dynamics with Finite Elements ............................................. 340
Dario Mangoni, Alessandro Tasora, and Simone Benatti

A Fast Explicit Integrator for Numerical Simulation of Multibody System Dynamics ............................................. 348
Hui Ren and Ping Zhou
Contents

Optimization and Sensitivity Analysis

The Discrete Hamiltonian-Based Adjoint Method for Some Optimization Problems in Multibody Dynamics ....................... 359
Paweł Maciąg, Paweł Malczyk, and Janusz Frączek

Dynamic Parameters Optimization and Identification of a Parallel Robot .................................................. 367
Taha Houda, Ali Amouri, Lotfi Beji, and Malik Mallem

Partial Shaking Force Balancing of 3-RRR Parallel Manipulators by Optimal Acceleration Control of the Total Center of Mass ........... 375
Jing Geng and Vigen Arakelian

Energy Expenditure Minimization for a Delta-2 Robot Through a Mixed Approach .................................................. 383
Giovanni Carabin, Ilaria Palomba, Erich Wehrle, and Renato Vidoni

Training a Four Legged Robot via Deep Reinforcement Learning and Multibody Simulation ........................................ 391
Simone Benatti, Alessandro Tasora, and Dario Mangoni

Efficient Simulation and Real-Time Applications

Two General Index-3 Semi-Recursive Formulations for the Dynamics of Multibody Systems ................................. 401
Daniel Dopico Dopico, Álvaro López Varela, and Alberto Luaces Fernández

Real-Time Capable Calculation of Reaction Forces of Multibody Systems Using Optimized Bushings on the Example of a Vehicle Wheel Suspension .................................................. 409
Frédéric Etienne Kracht and Dieter Schramm

A Machine Learning Approach for Minimal Coordinate Multibody Simulation .................................................. 417
Andrea Angeli, Frank Naets, and Wim Desmet

Efficient Particle Simulation Using a Two-Phase DEM-Lookup Approach .................................................. 425
Jonathan Jahnke, Stefan Steidel, Michael Burger, and Bernd Simeon

DARTS - Multibody Modeling, Simulation and Analysis Software ........ 433
Abhinandan Jain

Applications in Vehicle Dynamics and Aerospace Devices

Optimization of Geometric Parameters and Stiffness of Multi-Universal-Joint Drive Shaft Considering the Dynamics of Driveline .... 445
Xingyang Lu, Tongli Lu, and Jianwu Zhang
### Application of Multibody Dynamics in the Modelling of a Limited-Slip Differential

Michal Hajžman, Radek Bulín, and Štěpán Dyk

- Pages: 454

### Lateral Dynamics of Vehicles on a “Steerable” Roller Test Stand

Thomas Tentrup, Burkhard Corves, Jörg Neumann, Werner Krass, Jan-Lukas Archut, and Jascha Norman Paris

- Pages: 463

### Dynamic Interaction of Heavy Duty Vehicles and Expansion Joints

Daniel Rill, Christiane Butz, and Georg Rill

- Pages: 471

### A Study on the Behaviour of the Rotating Disk with the Damage on the Tread

Yasutaka Maki and Yoshiaki Terumichi

- Pages: 479

### Multibody Dynamics Analysis of Railway Vehicle with Independently Rotating Wheels Using Negative Tread Conicity

Yu Wang, Shihpin Lin, Hiroshi Tajima, and Yoshihiro Suda

- Pages: 487

### A Full-Vehicle Motion Simulator for Railways Applications

Roshan Pradhan, Vishnu Sukumar, Subir Kumar Saha, and Santosh Kumar Singh

- Pages: 495

### Simulation of the Maglev Train Transrapid Traveling on a Flexible Guideway Using the Multibody Systems Approach

Georg Schneider, Xin Liang, Florian Dignath, and Peter Eberhard

- Pages: 503

### Omni-Vehicle Dynamical Models Mutual Matching for Different Roller–Floor Contact Models

Kirill V. Gerasimov, Alexandra A. Zobova, and Ivan I. Kosenko

- Pages: 511

### Adjustment of Non-Holonomic Constraints by Absolutely Inelastic Tangent Impact in the Dynamics of an Omni-Vehicle

Alexandra A. Zobova, Kirill V. Gerasimov, and Ivan I. Kosenko

- Pages: 518

### Multibody Models and Simulations to Assess the Stability of Counterbalance Forklift Trucks

Michele Gardella and Alberto Martini

- Pages: 526

### Automatic Differentiation in Multibody Helicopter Simulation

Max Kontak, Melven Röhrig-Zöllner, Johannes Hofmann, and Felix Weiß

- Pages: 534

### Author Index

- Pages: 543
Biomechanics
Investigation of Tympanic Membrane Influences on Middle-Ear Impedance Measurements and Simulations

Benjamin Sackmann1,4, Birthe Warnholtz2, Jae Hoon Sim2, Dmitrii Burovikhin1, Ernst Dalhoff3, Peter Eberhard4, and Michael Lauxmann1

1 Reutlingen University, Alteburgstrasse 150, 72762 Reutlingen, Germany
benjamin.sackmann@reutlingen-university.de
2 University Hospital Zurich, Frauenklinikstrasse 24, 8091 Zurich, Switzerland
3 University of Tübingen, Elfriede-Aulhorn-Str. 5, 72076 Tübingen, Germany
4 Institute of Engineering and Computational Mechanics, University of Stuttgart, Pfaffenwaldring 9, 70569 Stuttgart, Germany

Abstract. This study simulates acoustic impedance measurements in the human ear canal and investigates error influences due to improperly accounted evanescence in the probe’s near field, cross-section area changes, curvature of the ear canal, and pressure inhomogeneities across the tympanic membrane, which arise mainly at frequencies above 10 kHz. Evanescence results from strongly damped modes of higher order, which can only be found in the near field of the sound source and are excited due to sharp cross-sectional changes as they occur at the transition from the probe loudspeaker to the ear canal. This means that different impedances are measured depending on the probe design. The influence of evanescence cannot be eliminated completely from measurements, however, it can be reduced by a probe design with larger distance between speaker and microphone. A completely different approach to account for the influence of evanescence is to evaluate impedance measurements with the help of a finite element model, which takes the precise arrangement of microphone and speaker in the measurement into account. The latter is shown in this study exemplary on impedance measurements at a tube terminated with a steel plate. Furthermore, the influences of shape changes of the tympanic membrane and ear canal curvature on impedance are investigated.

1 Introduction

Today’s audometric methods for the detection of hearing loss are mostly based on a comparison of measurements with standard curves representing the statistical range of normal hearing. Therefore, currently available clinical procedures for acoustic impedance measurements are often limited to a qualitative diagnosis of middle-ear pathologies with limited specificity. To resolve these limits, we previously established a model-based approach to evaluate the impedance measurements on individuals with a finite element (FE) model of the ear [5].
Currently there is no standardized commercial device available for measuring acoustic impedances in the ear canal (EC). Instead, only energy absorbance is measured taking into account only amplitude, but not phase information. The main reason for this is the dependency of the measured impedances on the geometry of the EC and on the probe design, which directly influences the evanescent waves. Evanescence results from strongly damped modes of higher order, which can only be found in the near field of the sound source and are excited due to sharp cross-sectional changes as they occur at the transition from the probe speaker to the EC.

Furthermore, the middle-ear impedance which describes the averaged velocity of the tympanic membrane (TM) caused by the pressure at the TM, is derived from the EC impedance based on the assumption that the residual EC can be modelled as an 1D-transmission line neglecting spatial pressure distribution and complex vibrations of the TM, which is questioned within this study.

For the investigations, a customized impedance probe is used as shown in Fig. 1 and it is shown how it can be modelled using an FE approach to account for the important influences such as evanescence. The investigations are done first on tubes with rigid termination and a flexible steel plate on which the impedance is measured. It is shown, how the parameters needed to simulate the measurements of the tube with steel plate termination can be achieved by first identifying the damping parameters of air in rigid tubes and then fitting the other parameters in a second step. Furthermore, the need to include evanescence in the simulations is studied. Furthermore, with an FE model of the ear, validated by comparing to measurements from literature, specific changes in the TM and EC shape are introduced and it is investigated how they affect the impedance.

2 Methods

2.1 How Is Impedance Measured?

The acoustic impedance $Z_{EC}$ is defined as the ratio of sound pressure $p$ and sound flow $u = vA_{isp}$ with the velocity $v$ and cross-section $A_{isp}$ of the driving loudspeaker and is measured at the EC entrance with a probe consisting of a microphone (EK-3024, Knowles) and loudspeaker (ER-3A, Etymotics), as shown in Fig. 1.

The probe is characterized by the two parameters probe impedance and probe pressure with a calibration procedure based on the Thévenin equivalent. These parameters are determined from pressure measurements on four waveguides of known impedance by solving an overdetermined system of equations using a least-squares method [1]. To improve the quality of the calibration, an error metric [6] is minimized by iteratively adjusting the calibration lengths.

In impedance measurements in the ear canal, the sound source has a smaller diameter than the measured tube or EC, which produces a sound field composed of propagating plane waves and rapidly decreasing evanescent waves as seen in Fig. 2.
Because the microphone is often placed in the plane of the loudspeaker, measurements and calibrations are affected by the evanescent waves. Currently this effect can be eliminated or reduced from the source parameters during the calibration by bevelling the probe to increase the aperture [7], as also done here, or by improved error metrics [4]. However, even though the Thévenin parameters of the probe can be corrected, the actual measurement at the ear is still impaired by evanescent modes and, therefore, depends on the specific probe [4].

The termination impedance $Z_{TM} = p_{TM}/u_{TM}$ is often calculated from sound pressure $p_{EC}$ and sound flow $u_{EC}$ at the probe in the EC assuming a plane wave propagation with the transmission matrix as

$$
\begin{bmatrix}
p_{TM} \\
u_{TM}
\end{bmatrix} = 
\begin{bmatrix}
a_{11} & a_{12} \\
a_{21} & a_{22}
\end{bmatrix} \cdot 
\begin{bmatrix}
p_{EC} \\
u_{EC}
\end{bmatrix} = 
\begin{bmatrix}
\cosh(jkL) & -Z_c\sinh(jkL) \\
-Z_c^{-1}\sinh(jkL) & \cosh(jkL)
\end{bmatrix} \cdot 
\begin{bmatrix}
p_{EC} \\
u_{EC}
\end{bmatrix}
$$

(1)

with the propagation factor $k$, the distance $L$ between probe and termination and the characteristic impedance $Z_c = \rho c/A$ (density $\rho$, speed of sound $c$, probe cross-section $A$). The positive wave propagation direction is defined from probe input to termination.
2.2 How Can the Measurements Be Simulated?

The impedance for the rigidly terminated calibration tubes and tubes with steel plate termination are derived from simulations with an FE code in Hyperworks (Altair Engineering Inc.). For the FE simulations, the air is modelled with acoustic FE using a pressure-based Eulerian approach, meaning that each node has a pressure degree of freedom. As the viscosity of air is very low in relation to the tube dimensions, a Rayleigh damping approximation is used for the air. In order to show the influence of evanescence on the simulated measurements, both a uniform excitation and a local excitation are simulated with the FE model. The loudspeaker is modelled as an ideal sound source with a constant frequency spectrum via a velocity boundary condition at the position of the probe. For uniform excitation the velocity $v$ is chosen to be $1 \cdot 10^{-5}$ m/s and the cylindrical diameter of the speaker is identical to the EC or tube diameter of 7 or 9 mm. For local excitation $v$ is chosen to be $2 \cdot 10^{-3}$ m/s and thus is adapted to the excitation area of the loudspeaker $A_{lsp}$ with a cylindrical diameter of 1 mm.

2.3 Simulation of Middle-Ear Impedance

An overview of the FE model used for the ear simulations is given on the right of Fig. 1. Details are thoroughly described in [2,3]. The geometry of the EC, tympanic cavity, TM, and middle-ear bones are reconstructed from computer tomography of a post-mortem temporal bone. The TM is modelled with shell elements and subdivided into six regions each with a constant thickness. The middle-ear bones are modelled as rigid bodies characterized by their mass and inertia. The ligaments, tendons and joints are represented by passive $6 \times 6$ spring-damper elements. For the simulation of the ear impedance, we use a uniform velocity excitation at the EC, which corresponds to an ideal probe-independent excitation and, thus, can be compared to other simulations. The length of the EC is 8 mm analogous to temporal bone measurements. For comparability of the simulations with literature measurements, the volume of the middle-ear cavity and the mastoid are selected according to an individual temporal bone (24L in [8]) to 0.9 cm$^3$ each, with the mastoid volume being considerably smaller compared to its natural state due to preparation.

3 Results and Discussion

3.1 Tubes, Measurements, and Simulations

By fitting the FE simulations to input-impedance measurements on tubes with rigid termination the Rayleigh damping coefficients for the air ($\alpha_{fluid} = 80$ Hz, $\beta_{fluid} = 2.55 \cdot 10^{-6}$ s) are derived. The assumption of fluid damping with Rayleigh coefficients in the simulation turned out to be suitable for prediction of the measured input impedances. For the probes with steel plate (diameter 9 mm) the damping parameters of air determined from the rigid tubes are used. Additionally, the parameters for the tube length (17.8 mm), plate thickness
(32.5 µm), Young’s Modulus (200 GPa), and the Rayleigh damping for the plate ($\alpha_{\text{structure}} = 16$ Hz, $\beta_{\text{structure}} = 1.3 \cdot 10^{-7}$ s) are determined from impedance measurements by parameter fitting. The impedance simulations of the steel-plate probes (green curve in Fig. 3) are in a very good agreement with the measurements. This can only be reached when the precise geometric arrangement of speaker and microphone is taken into account in the modelling.

![Graph](image)

**Fig. 3.** Comparison of measurements and simulations at the tube with steel plate termination. The simulation is done both with uniform as well as with local excitation. The green curve shows the impedance for local excitation with corrected microphone position according to the real probe.

Evanescence in the probe’s near field has an effect independent of the absolute pressure amplitude, in particular at the pressure anti-resonances where amplitude is small, while the resonances are hardly influenced. Evanescence lengthens the wavelength artificially and the anti-resonances shift to lower frequencies, which increases the frequency spacing to the uninfluenced resonances. A corresponding opposite effect is observed when the microphone is not located in the probe plane but extends axially further into the cavity, see Fig. 3. Therefore, in the model the precise geometric arrangement of microphone and loudspeaker needs to be included.

### 3.2 Ear Simulations

Figure 4 shows the pressure distribution in the EC, derived from an FE model with an EC length of 20 mm, similar to real patient measurements. As seen in the color plot, the EC can be divided into three regions of error sources.

In the probe’s near field (region I), evanescent modes occur and lead to pressure variations up to 30 dB SPL at 20 kHz at the probe side of the EC and to more than 20 dB SPL at 4 kHz, see Fig. 2, which is far away from useful tolerances. If the evanescence is not accounted for appropriately with consideration
of the exact position of the microphone, this will lead to large errors in the evaluation of impedance measurements, especially when the microphone is placed close to the speaker where small position variations of the microphone lead to completely different measured pressures. The probe should be designed in a way that the microphone is out of the near field, otherwise manufacturing tolerances of the probe might be crucial. There are two ways to design a probe, one which is more robust to tolerances, by increasing the distance between microphone and loudspeaker in the probe plane as much as possible, the other by moving forward the microphone into the measured EC. By using such a probe, the evanescence can be appropriately simulated with an FE model, and be accounted for in the evaluation of measurements.

In the wave propagation region II, errors are mainly introduced by an inappropriately accounted or not accounted cross-section area change in the EC and at higher frequencies (18–20 kHz) also by changes in the EC axis direction. Not accounting for the area change leads to tolerable errors below 4 kHz, however, large errors of more than 5 dB occur above 4 kHz in the case with conical EC, where the diameter of the EC at the probe is increased by 20% over a length of 8 mm, see Fig. 5. Therefore, the area function of the EC needs to be accounted for in the simulation. There are approaches available to estimate this function from measurements, however, further microphones for multiple pressure measurements in the EC are needed.

In the TM near field (region III) significant errors can be introduced by the large pressure inhomogeneity across the TM and leads to considerable errors of more than 5 dB at 10 kHz and even more than 10 dB above 15 kHz or 20 dB at 20 kHz as seen in Fig. 4 (right). This error type is mainly influenced by the angle between TM and EC axis because it originates from differences of the sound path lengths. These pressure inhomogeneities can be accounted for by an FE simulation, however, a simple 1D transmission-line model cannot account for them and will lead to large errors in the calculation of $Z_{TM}$ above 10 kHz.

To validate the model, two different cases are looked at, a normal case and a case with completely removed TM. Figure 6 shows the simulated $Z_{TM}$ compared to individual and averaged measurements from literature [8]. Compared to the literature data, the model shows a valid modelling of the air in the cavities and also the curves of the normal ear match the important characteristics in
the literature data qualitatively well. In some frequency ranges the simulated impedance has magnitudes up to a factor of 2 times lower than those of temporal bone 24L, however, changes in that magnitudes are within the confidence interval of normal ears.

Fig. 5. TM influences on impedance compared to EC influences.

For an evaluation of impedance measurements on individuals with the FE model of the ear it is important to know furthermore the effects of specific shape and thickness changes of TM. Therefore, sensitivity studies seen in Fig. 5 are conducted.

Changes in the TM thickness distribution lead to significant impedance changes at about 2 and 3.5 kHz. Changes in the curvature of the TM lead to significant impedance changes from 1 to 4 kHz. A flat TM leads to changes up to 10 dB at 1 kHz and significantly changes the static stiffness. Considering the exact geometry of the TM in an FE simulation based evaluation of impedance measurements is therefore essential. Main influences of the inclination angle between EC and TM, which was changed from perpendicular to 45° are above 8 kHz and of less importance.

Fig. 6. Simulation of TM impedance with the FE model of the ear for a temporal bone with sealed antrum compared to literature data [8] of a normal ear and an ear with removed TM.
4 Conclusion

This study investigated how impedance measurements can be simulated and how to account for error sources in measurements. While the inclination angle of the EC to the TM is of less importance, however, geometry changes of the TM lead to large differences in impedance. Therefore, the TM geometry should be accounted for with optical coherence tomography when simulating impedance measurements. Evanescence leads to considerable errors even at 4 kHz, and, therefore, the exact geometric arrangement of microphone and loudspeaker needs to be included in the FE model and must be accounted for in order to obtain accurate impedance measurements. To be able to account for the pressure inhomogeneities above 10 kHz FE simulation is indispensable. The area function of the EC needs to be accounted for above 4 kHz otherwise in temporal bone preparations a straight EC should be used.

Acknowledgements. This work has been funded by Volkswagen Foundation (Az. 93949) and a scholarship of the Ministry of Science, Research and Art Baden-Württemberg (MWK). This support is gratefully acknowledged.

References

Anterior Cruciate Ligament Injuries Alter the Kinematics and Kinetics of Knees with or Without Meniscal Deficiency

Xiaode Liu¹, Hongshi Huang², Shuang Ren², Yingfang Ao²(✉), and Qiguo Rong¹(✉)

¹ Department of Mechanics and Engineering Science, College of Engineering, Peking University, Beijing 100871, China
qrong@pku.edu.cn
² Institute of Sports Medicine, Peking University Third Hospital, Beijing 100191, China
aoyingfang@163.com

Abstract. The purpose of this paper was to study the biomechanical behaviors of knees with anterior cruciate ligament deficient (ACLD) with or without a combined medial or/and lateral meniscal injury during level walking. The motion capture system and the modeling system (AnyBody) were applied to simulate the kinematic and kinetic properties of knees. The results show that the knees with ACLD exhibited significantly less extension than the control knees at the mid stance. A lower extension moment and adduction moment in all ACLD-affected knees were detected during the terminal stance when compared with control knees. The ACLDML group showed significantly lower proximodistal compressive forces and anteroposterior and mediolateral shear forces, while the shear forces tended to increase in the ACLD, ACLDL, and ACLDM groups.

1 Introduction

It is well known that the anterior cruciate ligament (ACL) plays an important role in knee joint stability by limiting anterior tibial translation and maintaining axial and transverse rotation of the knee [1]. Many studies have investigated ACL biomechanics under various loading conditions, especially for the patients who have ACL injuries. ACL deficient (ACLD) knees tend to exhibit abnormal biomechanical characteristics compared to the control group and often associate with an incidence of knee osteoarthritis. Majority of previous studies have investigated the characteristics of knees with isolated ACLD. ACL injuries, however, are commonly combined with meniscal tears. In the United States, approximately 40% to 80% of ACL injuries are combined with menisci injuries [2]. Bellabarba et al. [3] reported that 41% to 82% of acute ACL-injured knees and 58% to 100% of knees with chronic ACL deficiency had meniscal tears. Some studies [4–6] have showed that the location of the meniscus tear could influence kinematics in ACLD knees.
However, the effect of different meniscus injury patterns on the kinematics and kinetics in ACLD knees during gait, has not been well studied. Meanwhile, accurate determination of kinetics involves however difficulties and limitations in both in vitro cadaver and in vivo imaging studies. Musculoskeletal modeling on the other could circumvent such shortcomings. However, there are few studies focusing on the musculoskeletal modeling analysis. In this study, based on a multi-body dynamics software, a three-dimensional musculoskeletal modeling method was used to investigate the kinematics and kinetics of ACLD knees with or without a combined medial or lateral meniscal injuries during level walking. We hypothesized that gait mechanics would present a more abnormal pattern in ACLD knees with medial or/and lateral meniscal injury than in those without.

2 Methods

Between January 2014 and December 2016, the experimental data was collected by using an optical motion capture system during normal walking. The ethical approval was obtained from the university’s ethics committee and written informed consent was attained from all participants. The ACLD knees were diagnosed by clinical examination, magnetic resonance imaging and confirmed during arthroscopic ACL reconstruction surgery. 29 patients with unilateral ACLD knees (contralateral side intact; injury time range: 6 months–4 years, age range: 18–34 years; body mass index rage, 20.52–35.65 kg/m$^2$) were recruited before undergoing ACL reconstruction. Among these patients, 12 patients had isolated unilateral ACL injuries (ACLD group), 5 had combined ACL and lateral meniscal injuries (ACLDL group), 5 had combined ACL and medial meniscal injuries (ACLDM group) and 7 had combined ACL and medial/lateral meniscal injuries (ACLDML group). When evaluating the meniscal tears, we did not consider the type of tear (i.e., longitudinal root tear, horizontal cleavage tear, or complex tear) because of the limited sample size. None of the knee cartilage defects were higher than grade II according to the Outerbridge system [7]. 15 healthy male with no history of musculoskeletal or neuromuscular disorders in the lower extremities and no measurable ligamentous instability on clinical examination volunteered for this study (Control group).

The experimental data were collected by 100-Hz eight-camera motion capture system (Vicon MX; Oxford Metrics, Yarnton, Oxfordshire, UK), and ground-reaction forces were measured using two 1000-Hz embedded force plates (Advanced Mechanical Technology Inc., Watertown, MA, USA). A set of markers were attached to participants lower limbs to track segmental motion during walking. Based on the validated plug-in-gait model, anatomical markers were taped to following anatomical lower limbs locations: the anterior and posterior superior iliac spine, medial and lateral femoral epicondyles, malleoli, and medial and lateral sides of the calcaneus, the frontal and lateral aspects of the thigh and the shank, posterior part of the calcaneus, heads of the first, second, and fifth metatarsal bones, base of the first metatarsal bone, and navicular, hallux.
After performing a standing trial, participants were asked to walk from a specified point. A successful gait trial was recorded when each foot stepped on the force plates at a self-selected speed. Five successful gait trials were assessed for each participant.

A three-dimensional (3D) musculoskeletal model was employed using multibody dynamics software (AnyBody Modeling System, version 6.0.5; AnyBody Technology A/S). This model was constructed based on the University of Twente lower extremity model (TLEM) and has been validated to calculate muscle forces and joint moments [8]. The model consists of 160 muscle-tendon actuators and 6 joint degrees of freedom: the hip joint was modeled as a spherical joint with 3 degrees of freedom; the knee joint was modeled as a hinge joint with only 1 degree of freedom for flexion-extension, and a universal joint was considered for the ankle subtalar complex.

As for the multi-body dynamics analysis in the TLEM, initial condition operations were firstly performed to identify the parameters of segment lengths and the (virtual) marker positions. Anatomic frames on the femur and tibia corresponding to the static standing trial were defined based on the respective tibiofemoral joint (TFJ) landmarks (Fig. 1). In order to match the outer markers (experimental data) well with the inner markers (markers attached to the TLEM model based on the same plug-in-gait model), least-squares minimization were performed between the outer markers and inner markers positions. The TLEM model was re-modeled according to our anthropometric data (body weight, body height, pelvis width, thigh, shanks, and foot length) for each subject, and the personalized model was scaled with a mass-fat scaling algorithm. Kinematics and inverse dynamics analysis were successively performed for each simulation: first, the skeletal model was employed to calculate the joint kinematic waveforms and segmental motions related to the experimental gait trials; second, the musculoskeletal model was used in the inverse dynamic simulations to calculate the joint forces and moments. Joint contact forces were derived from the inter-segment loading and muscle force that acted on the joint. For the muscle recruitment, we adopted a min/max recruitment solver in the software to solve the muscle redundancy problem. Five different gait trials were simulated for each participant, and the average values (including the sagittal joint angle, 3D reaction forces, and joint moments of the TFJ were compared between the control and ACLDs knees during level walking.

3 Results

The ACLD knees, with or without meniscal deficiency, had significant less extension (Fig. 2A) and lower extension moments (Fig. 2C) than the control knees at mid-stance, including the maximum extension moment (ACLD: $-0.011 \pm 0.014 \text{Nm/(kg*m)}$, ACLDL: $-0.011 \pm 0.010 \text{Nm/(kg*m)}$, ACLDM: $-0.012 \pm 0.011 \text{Nm/(kg*m)}$, ACLDML: $-0.018 \pm 0.008 \text{Nm/(kg*m)}$, control: $-0.024 \pm 0.017 \text{Nm/(kg*m)}$, $p < 0.05$). Compared with the control group, the ACLDL and ACLDML groups showed a decreased proximal-distal compressive
forces, while ACLDM group presented an increased trend. For group ACLDML, in particular, the reduction in the maximum compressive force was significant (ACLDML: 3.45 ± 0.52 N/kg, control: 4.46 ± 1.65 N/kg, p < 0.05) (Fig. 2B). Meanwhile, there is no significant statistical difference at the maximum flexion moment, except for the comparison between control knees and ACLDML knees (Fig. 2C). No significant differences in the flexion moment were observed between ACLDL and ACLDM groups during the stance phase.

4 Discussion

This study investigated the sagittal kinematics and 3D kinetics of ACLD-affected knees with or without meniscal injuries during level walking. The results of our current research support the initial hypothesis that the kinematic and kinetic response of ACLD-affected knees could be dependent on the type of meniscal injuries. Some previous studies have investigated the effects of meniscal tears on the gait parameters of ACLD-affected knees [4–6]. Ren et al. [4] showed that ACLD-affected knees with medial meniscal posterior horn tears exhibited extension deficiency, more external tibial rotation, and lower extension and internal rotation moments. Zhang et al. [6] recently reported that a combined
Fig. 2. TFJ kinematics and kinetics of control and patients (ACLD, ACLDL, ACLDM and ACLDML) knees. Segments with significant statistical differences between the patients and the control groups were marked with asterisks. The green shaded area represents means standard deviation of the control group. HS: heel strike; CHS: contralateral heel strike; TO: toe off; CTO: contralateral toe off. deg: degree.
ACL/meniscus injury could alter the kinematics of ACLD-affected knees compared with knees with isolated ACL injury. However, these studies did not assess the alterations in the joint contact force and moment in patients with combinations of ACLD and different types of meniscal tears (e.g., lateral or/and medial meniscal tears). As far as we know, this is the first inverse dynamical study to evaluate ACLD-affected knees with concomitant medial or/and lateral meniscal tears. Our result were in line with the results of previous studies [9] and in vivo measurements [10], and showed that meniscal injuries altered the kinematics and kinetics of the ACLD knees when compared with an isolated ACL injury. The less knee extension in ACLD knees throughout the stance phase may be explained by a protective adaptation strategy for the ACLD knees to avoid excessive anterior tibial displacement at maximum extension. As the injury knees result in a lower demand on the quadriceps muscles, the reduction in the maximum flexion of ACLD knees may result from the nervous system to avoid pain. Subjects with knee stability after ACL rupture consistently stabilize their knee with a stiffening strategy involving less knee motion and higher muscle contraction. Joint unloading may be associated with the cascade of early degenerative changes at the knee [11], so the lower compressive forces in ACLDML knees, with combined meniscal injuries, could result in the higher incidence of cartilage degenerations. The increased compressive forces in ACLD knees may be related to the type of medial meniscal injuries, as the menisci plays an important role in transmitting tibiofemoral loads and reducing pressure on articular cartilage [12] and release of the anterior intermeniscal ligament results in increased peak contact pressures in the medial compartment of the knee [13]. Chronic ACLD subjects tended to present a muscle contraction adaptation strategy, which increased the activity of quadriceps and hamstrings. Therefore, the lower extension in the study may be related to the muscle co-contraction adopted by ACLD knees to maintain knee stability.

5 Conclusion

The results indicate that a combined ACL/meniscal injuries could alter the kinematics and kinetics of ACLD knees depending on the presence and type of meniscal tears. The abnormal biomechanics behavior may be associated with the cascade of early degenerative changes and subsequent onset of osteoarthritis.

Acknowledgements. This study was supported by Beijing Municipal Natural Science Foundation (No. 7172120) and National Natural Science Foundation of China (No. 11872074).
References

Investigation of Inhomogeneous Stiffness and Damping Characteristics of the Human Stapedial Annular Ligament

D. Burovikhin1, Benjamin Sackmann1, Merlin Schär2, J. H. Sim2, P. Eberhard3, and M. Lauxmann1

1 Reutlingen University, Alteburgstr. 150, 72762 Reutlingen, Germany
{dmitrii.burovikhin,benjamin.sackmann,michael.lauxmann}@reutlingen-university.de
2 University Hospital Zürich, Frauenklinikstrasse 24, 8091 Zürich, Switzerland
{merlin.schaer,jaehoon.sim}@usz.ch
3 Institute of Engineering and Computational Mechanics, University of Stuttgart, Pfaffenwaldring 9, 70569 Stuttgart, Germany
peter.eberhard@itm.uni-stuttgart.de

Abstract. This study describes a non-contact measuring and system identification procedure for evaluating inhomogeneous stiffness and damping characteristics of the annular ligament in the physiological amplitude and frequency range without the application of large static external forces that can cause unnatural displacements of the stapes. To verify the procedure, measurements were first conducted on a steel beam. Then, measurements on an individual human cadaveric temporal bone sample were performed. The estimated results support the inhomogeneous stiffness and damping distribution of the annular ligament and are in a good agreement with the multiphoton microscopy results which show that the posterior-inferior corner of the stapes footplate is the stiffest region of the annular ligament.

1 Introduction

The stapes is a bone found in the middle ear of humans and other mammals which is involved in the conduction of sound vibrations to the inner ear, see Fig. 1. The stapedial annular ligament (AL) is a ring of fibrous tissue that connects the base of the stapes, its footplate, to the oval window of the inner ear. The anatomical dimensions of the AL support the hypothesis of an inhomogeneous stiffness distribution. In [3] it has been revealed that the cross-section of the AL is posteriorly narrower and thicker, resulting in a higher stiffness on the posterior side, and in [1] it is stated that the properties of the AL largely determine the transfer characteristic of the middle ear in the lower frequency range.

A number of studies on the topic of the stiffness characteristics of the human stapedial AL were conducted in the past, but all of them were focused on determining the stiffness properties of the AL in the quasi-static frequency range by...